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(54) COMPENSATING FOR FLUID CHANNEL DIMENSION VARIATIONS IN A FLOWMETER

(57) A flowmeter for measuring small differential fluid flow rates, of dialysate input to and output from a dialyser, comprises two similar, rectangular section channels 11, 12 and a compensating or calibrating circuit eliminating the effects of variations in the channel heights or "diameters". The channels are in a magnetic field H, so that the flow of dialysate generates an e.m.f. between electrodes 13 and 14 in the channels, which is representative of the differential flow rate. The compensating for channel height mismatch is effected by using a signal proportional to common mode flow rate from electrodes 15 and 16, which is amplified by a factor whose ratio to the differential flow rate signal amplification factor is proportional to channel height mismatch, and summing the two amplified signals. The common mode flow rate amplification factor is determinable by a calibration procedure. A suitable circuit is disclosed (Figs 6 and 7 not shown).

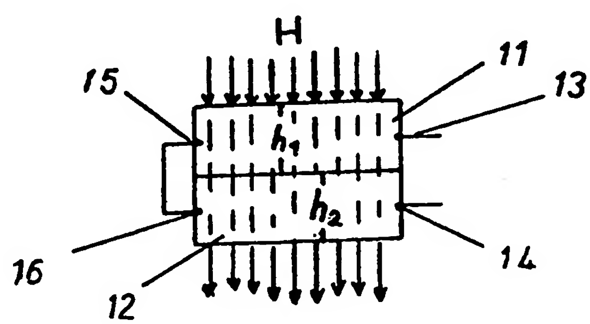

Fig 2

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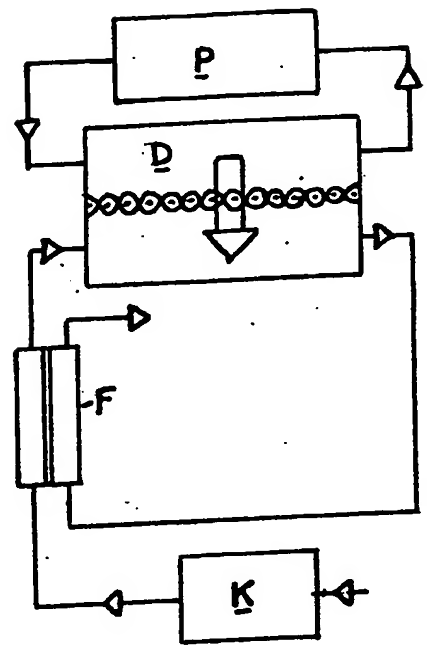


Fig 1

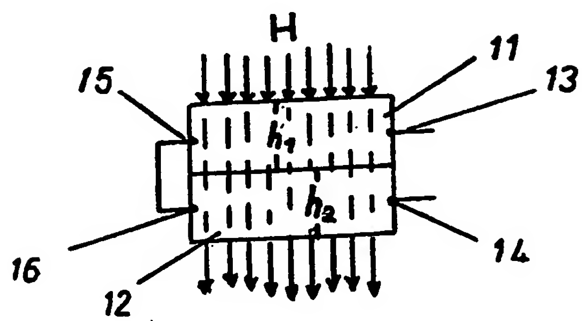


Fig 2

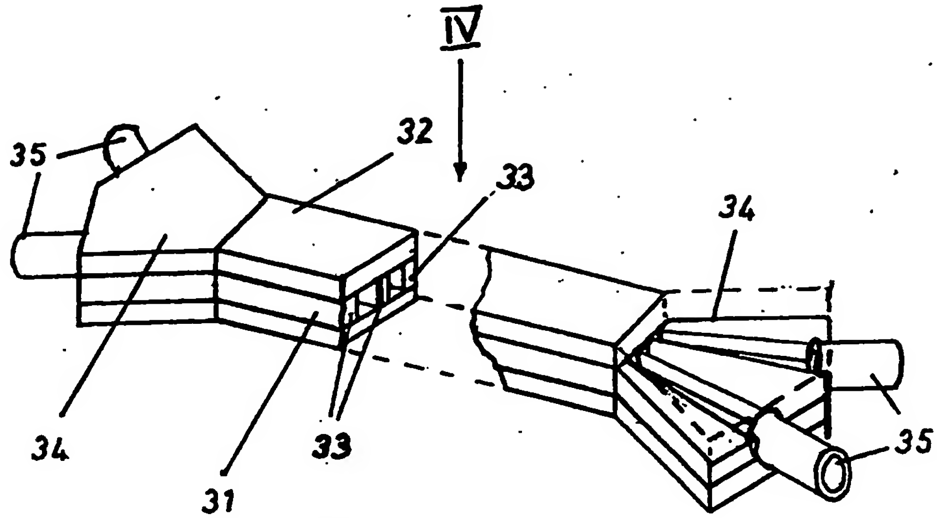
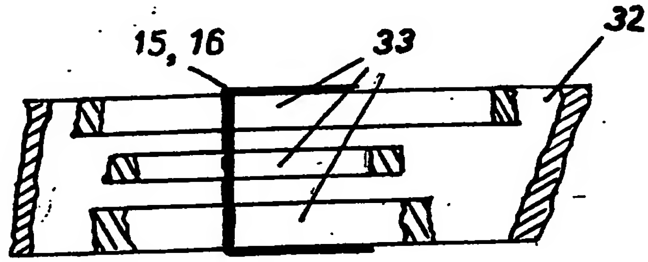


Fig 3

Fig 4



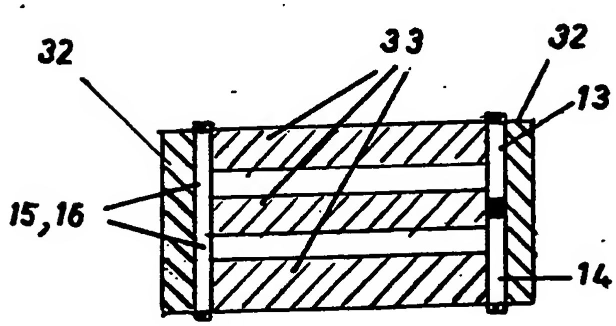


Fig 5

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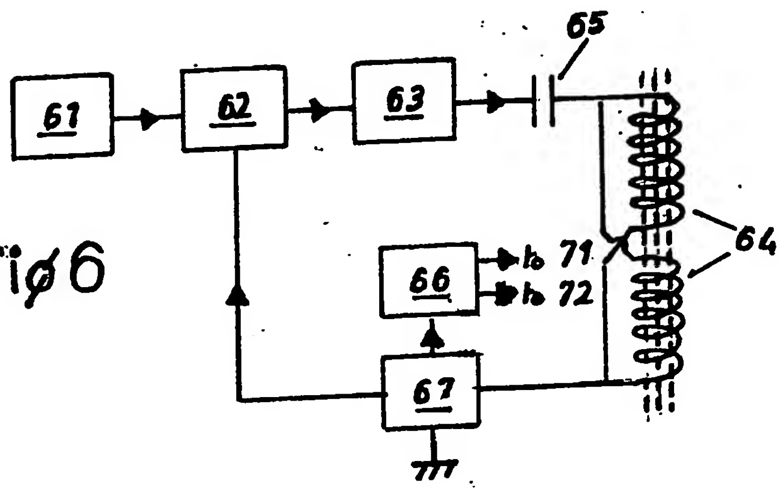

Fig 6

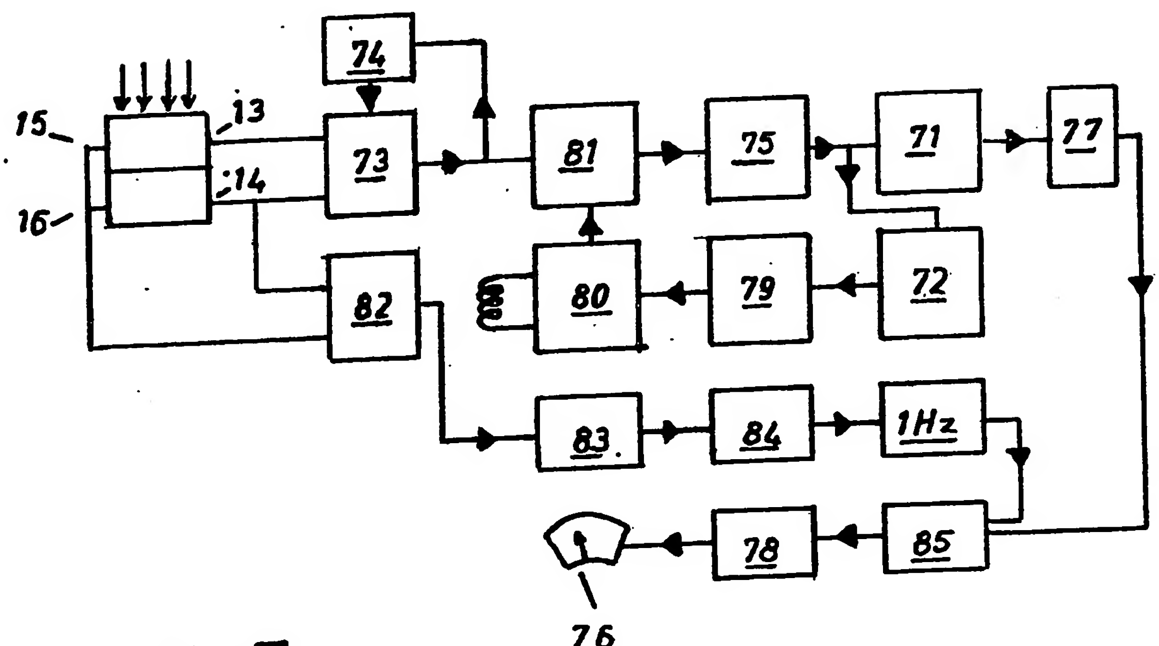

Fig 7

SPECIFICATION

MEASUREMENT METHOD AND APPARATUS

This invention relates to the measurement of small differential fluid flow rate, such as may be used in haemodialysis.

In Paper E-2 given at the Conference on Fluid Flow Measurement in the Mid 1970s, held at the Birniehill Institute, National Engineering Laboratory, East Kilbride, Glasgow on 8—10 April, 1975, entitled "Precision Differential Fluid Flow Measurement", J. O. Gray and M. L. Sanderson described a principle of measuring very small differential flow rates. The principle was primarily for use in dialysis, as a way of monitoring the rate of loss of water from the patient. The dialysate input to an output from the dialyser are passed in counter-current through adjacent, similar, rectangular section channels of a transducer. The channels are in a magnetic field, so that the flow of the dialysate generates an e.m.f. between electrodes in the channels. The electrodes are arranged so that the output e.m.f. from the transducer is representative of the differential flow rate.

In dialysis, the input flow rate of dialysate is usually in the region 400—600 ml/min, and the amount of water added to this flow in the dialyser—being the amount extracted from the patient—is of the order of a few ml/min.

The principle described by Gray and Sanderson is indeed sensitive to these very small differentials. However, it is not practicable to manufacture the transducer to the tolerances required to ensure reasonable accuracy in this measurement.

This invention provides a method and apparatus by which mismatch in channel dimensions in apparatus of this type (known hereinafter as the type referred to) can be compensated for. It turns out that provided the channel widths are reasonably well matched, the critical matching dimension is the channel height—that is to say, the dimension parallel to the magnetic field. Any mismatch in channel height is found to contribute to error in the output signal proportionally to common mode flow rate.

The invention comprises a method for compensating for channel height mismatch in the measurement of differential fluid flow rate in apparatus of the type referred to, in which a first signal proportional to the differential flow rate is amplified by a first amplification factor and a second signal proportional to the common mode flow rate is amplified by a second amplification factor whose ratio to the first said factor is equal to the fractional channel height mismatch, the two amplified signals being added together.

The said second amplification factor may be fixed by calibration.

The invention also comprises compensating apparatus for channel height mismatch in the measurement of differential fluid flow rate in apparatus of the type referred to in which a first signal proportional to the differential flow rate is amplified by a first amplification factor, comprising a compensating amplifier adapted to amplify a second signal proportional to common mode flow rate by a second amplification factor whose ratio to the first amplification factor is equal to the fractional channel height mismatch, and summing means adapted to add the two amplified signals together.

The said compensating amplifier is preferably adjustable for calibration.

One embodiment of compensating apparatus and its method of operation according to the invention will now be described with reference to the accompanying drawings in which:—

Figure 1 is a diagram showing haemodialysis being monitored by differential fluid flow rate measurement,

Figure 2 is a diagrammatic cross-section of a differential fluid flow rate transducer,

Figure 3 is a part cut away perspective view of a channel system for a transducer of the kind shown in Figure 2,

Figure 4 is a view along arrow IV of Figure 3, with part removed to expose the interior of the channel system,

Figure 5 is a cross-section of the channel system shown in Figure 3,

Figure 6 is a block diagram of a driver circuit producing the magnetic field in the transducer, and

Figure 7 is a block diagram of an electronic circuit amplifying the output signals from the transducer.

In conventional haemodialysis, illustrated in Figure 1, arterial blood from the patient P is passed through a dialyser such as a Kiil dialyser D along one side of a semipermeable membrane and returned to a vein in the patient. A dialysate fluid from a kidney machine (such as a Lucas kidney machine) K passes through the dialyser D on the other side of the membrane, and water and certain dissolved matter transfer across the membrane. A principle object of the dialysis is the removal of water from the blood, and this results in the dialysate output from the dialyser being greater than the dialysate input thereto.

A differential flow meter is the only effective way of monitoring this process, apart from repeated or continuous weighing which has obvious limitations. The rate of removal of water from the patient, at between 1 ml and 10 ml/min is too small for accurate measurement by ordinary flowmeters. Differential flow measurement is applied to haemodialysis by passing the dialysate flows into and out from the dialyser through a differential flowmeter F like that illustrated in Figure 2. The arrangement according to the invention is shown in Figure 1. According to the principle described by Gray and Sanderson, the two flows travelled in opposite directions through the flowmeter, whereas in the practical method now described, they travel in the same direction, eliminating the large effect due to the common

mode flow (i.e., the input flow) and leaving only the small effect of the differential flow (i.e., the net flow across the membrane.) The measurement is applied to the dialysate flow rather than to the blood flow for clinical reasons.

The flowmeter F comprises identical channels 11, 12 of rectangular cross-section. A magnetic field H is applied by an electrically energised stack of laminations (not shown) so as to be uniform at a flux density of about 1500 lines/cm^2 over the region. Electrodes 13, 14, 15, 16 are provided in contact with the fluid in the channels 11, 12, electrodes 15, 16 being connected together. The dialysate being ionic and therefore a conductor, its movement through the field generates e.m.f.s between the electrodes.

The e.m.f. between electrodes 13, 14 can be shown to be

$$e=B(Q_1/h_1-Q_2/h_2)\times 10^{-8} \text{ volts}$$

where B is the common flux density in lines/cm^2, h_1, h_2 are the heights of the channels 11, 12 in cms and Q_1, Q_2 are the quantities of fluid flowing in the channels 11, 12 in ml/min.

If, as Gray and Sanderson assume, $h_1=h_2=h$, then we have

$$e=B/h(Q_1-Q_2)\times 10^{-8} \text{ volts}$$

However, for this e.m.f. to be easily measureable, and to ensure a uniform magnetic flux throughout the measurement region with a reasonable coil size, it is desirable for h to be small, and this means that in practice it is difficult to ensure that h_1 and h_2 are sufficiently closely matched.

The embodiment of a channel system illustrated in Figures 3, 4 and 5 is adapted for precise engineering to close tolerances. The channel system comprises a crown glass or borosilicate body 31 built up of accurately ground and polished side plates 32 and middle pieces 33. Platinum electrodes 13, 14, 15, 16 are set in grooves cut into the side plates 32, the grooves extending to the edges of the plates so the electrodes can be taken out and laid along the outside of the body, as shown in Figure 4. The ends 34 of the body are flared and bored to accept tubular connectors 35. The manner of construction and the materials used must satisfy clinical requirements—in particular, they must be free from toxin and be capable of cleaning and sterilisation—and an assembly of glass plates cemented or preferably fused together is ideal in this regard and also allows very accurate construction by methods used in the manufacture of optical glasses.

A convenient nominal height for each of the channels 11, 12 is 1.6 mm. However, the best practical tolerance for matching the channel heights is then 0.0025 mm. This is not good enough for acceptably accurate flow rate measurement over the range of common mode flow rates experienced in haemodialysis.

Let us say that

$$Q_1-Q_2=\Delta Q$$

and

$$h_1-h_2=\Delta h$$

then

$$e=B(\Delta Q/h_1-Q_2\Delta h/h_2^2)\times 10^{-8}$$

The second term in brackets, in Q_2 and Δh, shows that any channel height mismatch affects the differential e.m.f. proportionally to the flow rate.

For the height and tolerance given above and assuming a differential flow rate of 1 ml/min and a common mode flow rate of 500 ml/min, we have

$$\Delta Q/h_1=0.625 \text{ and } Q_2\Delta h/h_2^2=0.488 \text{ (max)}$$

Thus the height mismatch can introduce error into the observed e.m.f. of as much as ±78%. According to the invention, this error is eliminated by deriving a compensating e.m.f. from the transducer, suitably amplifying this compensating e.m.f. and adding it to the amplified differential e.m.f. to produce an e.m.f. independent of the common mode flow Q_2.

Now the e.m.f. between electrodes 14 and 16 is given by

$$e_{14,16}=BQ_2/h_2\times 10^{-8} \text{ volts}$$

If we amplify e and $e_{14,16}$ by different amplification factors F_1, and F_2 respectively, then

$$E=F_1e+F_2e_{14,16}$$

is independent of Q_2 if

$$F_2/F_1=\Delta h/h_2$$

As a rule, F_1 will be fixed and F_2 adjustable for initial calibration of a particular electronic signal processing arrangement to a particular transducer.

Figures 6 and 7 illustrate the electronic equipment for operating the flowmeter. Figure 6 shows the drive circuit for energising the electromagnet. The energising current is alternating at 80 Hz to avoid polarisation effects at the electrodes. The 80 Hz signal derived by dividing a 10 MHz signal from a crystal oscillator 61, is passed via a voltage controlled attenuator 62 to an amplifier 63 and to the transducer coils 64 which are in series with a resonating capacitor 65. The coil current is monitored by a current sensing assembly 67 which feeds back a rectified and integrated signal to control the attenuator 62 to ensure amplitude stability. This way of generating the excitation current avoids the use of a tuned amplifier, which has an intrinsically unstable 90° phase shift.

The transducer coil energising current is also passed through phase shifting circuits 66 to orthogonal phase sensitive detectors 71, 72

(Figure 7). These circuits 66 comprise a phase locked loop with frequency multiplication and a digitally derived 90° phase shift, in preference to a simple analogue phase shift, which is subject to temperature instability.

The differential e.m.f. from electrodes 13, 14 of the transducer is amplified in a pre-amplifier 73 with an automatic d.c. balance 74 to cancel any d.c. changes produced by temperature changes or amplifier drift. The signal, further amplifier in an a.c. amplifier 75 which eliminates any remaining d.c. offsets is output through the phase sensitive detector 71, which detects signals in phase with the transducer coil current, and fed to a meter 76 (or a chart recorder, not shown) via 1 Hz and 10 second filters 71, 78 respectively. These filters improve transient and general response times.

The second phase sensitive detector 72 detects signals out of phase with the energising current. This out of phase signal is integrated in an integrating circuit 79 and fed to a quadrature suppression circuit 80 which controls the magnitude of a quadrature signal obtained from a winding of the transducer energising coil. The output of the suppression circuit 80 is added to the pre-amplified differential e.m.f. in a summing circuit 81.

The common mode signal from electrodes 14, 16 is pre-amplified in an amplifier 82, and further amplified in an a.c. amplifier 83, the inphase output of which is detected in the phase sensitive detector 84 and fed via a 1 Hz filter to summing circuit 85, wherein it is added to the main signal. One or other of amplifiers 82, 83 is adjustable so that when the electronics and the transducer are first brought together, the arrangement can be calibrated. This can be done by passing the same flow through both channels—simply connecting the output of one to the input of the other—and adjusting the gain until a zero reading is obtained. The instrument is then ready for use in dialysis to give a compensated reading independent of the common mode flow rate.

The apparatus is preferably arranged in two separately cased units, so that the bulk of the electronics is contained away from the transducer, where leakage of dialysate could harm both the patient and the electronics. For additional protection for the patient, and to reduce d.c. signal generation due to polarisation, all signals from the transducer are capacitively coupled into the electronic circuits. The pre-amplifiers 73 and 82, however, are best included in the transducer arrangement, so that small e.m.f.s are not attenuated by lead resistances.

It is assumed that suitable attention will be paid to the stability of circuits used in the electronics, and that provision will be made for eliminating mains fluctuations, it being remembered that dialysis extends over several hours.

Of course, the invention is not limited to use in dialysis. It may find application in many situations requiring measurement of very small differential flow rates, such, for example, as monitoring the mixing of fluids, measuring rates of evaporation and so on.

CLAIMS

1. A method for compensating for channel height mismatch in the measurement of differential fluid flow rate in apparatus of the type referred to, characterised in that a first signal proportional to the differential flow rate is amplified by a first amplification factor and a second signal proportional to the common mode flow rate is amplified by a second amplification factor whose ratio to the first said factor is equal to the fractional channel height mismatch, the two amplified signals being added together.

2. A method according to claim 1, characterised in that the said second amplification factor is fixed by calibration.

3. Compensating apparatus for channel height mismatch in the measurement of differential fluid flow rate in apparatus of the type referred to in which a first signal proportional to the differential flow rate is amplified by a first amplification factor, characterised by comprising a compensating amplifier adapted to amplify a second signal proportional to common mode flow rate by a second amplification factor whose ratio to the first amplification factor is equal to the fractional channel height mismatch, and summing means adapted to add the two amplified signals together.

4. Apparatus according to claim 3, characterised in that the said compensating amplifier is adjustable for calibration.

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